Focused high frequency needle transducer for ultrasonic imaging and trapping

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A miniature focused needle transducer (<1 mm) was fabricated using the press-focusing technique. The measured pulse-echo waveform showed the transducer had center frequency of 57.5 MHz with 54% bandwidth and 14 dB insertion loss. To evaluate the performance of this type of transducer, *in vitro* ultrasonic biomicroscopy imaging on the rabbit eye was obtained. Moreover, a single beam acoustic trapping experiment was performed using this transducer. Trapping of targeted particle size smaller than the ultrasonic wavelength was observed. Potential applications of these devices include minimally invasive measurements of retinal blood flow and single beam acoustic trapping of microparticles. © *2012 American Institute of Physics*. [http://dx.doi.org/10.1063/1.4736731]

Recent advances in high frequency ultrasonic transducer have made further applications such as high frequency imaging and acoustic microparticle trapping possible.[1,2] Higher center frequency can achieve better spatial resolution of ultrasonic imaging as well as smaller particle manipulation. For a number of applications, the size of the transducer is crucial and there is a need to miniaturize the transducer size. A small diameter unfocused needle transducer has been reported for ultrasonic imaging and blood flow measurements in the eye.[3] In this paper, focused needle transducers that will achieve better spatial resolution than unfocused transducers will be reported. In addition, a focused needle ultrasonic transducer offers more flexibility in carrying out single beam acoustic trapping experiments given the extremely congested environment in the small area within which the experiments are performed.[4]

For a miniaturized transducer, piezoelectric materials with high dielectric constant are more desirable since the electrical impedance of a transducer is inversely proportional to the dielectric constant of the piezoelectric material.[5] Comparing with other materials, $Pb(Mg_{1/3}Nb_{2/3})O_3$-$PbTiO_3$ (PMN-PT) single crystal is a promising candidate to build small transducers because of its high dielectric constant (1000–5000). Additionally, it has higher piezoelectric mechanical coupling coefficient (0.58), which enhances energy conversion and improves the sensitivity of the transducer.[6] Traditionally, two types of focused transducers can be fabricated via lens-focusing and press-focusing (self-focusing) techniques. Comparing with lens-focused transducers, self-focused transducers that typically yield higher sensitivity and lower *f*-number ($f_{\#}$) can be fabricated quicker and the devices produced are more consistent.[5] Hence, PMN-PT single crystal self-focused needle transducers were developed in this study to meet the demands of further applications.

Fig. 1 shows the cross-sectional view of a PMN-PT self-focused needle transducer. The fabrication process is described as follows. First, the PMN-PT single crystal was lapped to 23 μm, and then Cr/Au (50/100 nm) electrodes were sputtered on both sides of the PMN-PT. The silver epoxy matching layer was cast on the PMN-PT and lapped to 7 μm. E-solder 3022 was cast onto the opposite side as the backing material and lapped to 2 mm. The sample was then diced to 0.4×0.4 mm^2 posts and housed inside a polyimide tube with an inner diameter of 0.57 mm. A lead wire was connected to the backing layer with an additional amount of conductive epoxy. The polyimide tube provided electrical isolation from the 20-gauge needle housing with an outer and inner diameter of 0.9 mm and 0.7 mm, respectively. The device was press focused by a steel ball bearing at 75 °C to obtain a 0.4 mm focus and an *f*-number of 1. Next, a layer of Cr/Au was sputtered across the transducer face to form the ground plane connection. A 4 μm parylene layer was vapor-deposited on the front face of the transducer, serving as an acoustic matching layer and a protection layer. Last, the transducer was housed in modified SMA connector. The measured pulse-echo waveform with normalized frequency spectrum and lateral beam profile are given in Fig. 2. The center frequency was 57.5 MHz with a −6 dB bandwidth of 54% and the insertion loss 14 dB. The −6 dB beam width was determined to be 24 μm by scanning a 6 μm diameter tungsten wire as pulse-echo target at the focus, where translational scans at the focus of the transducer yield the lateral profile of projections of the beam.[7] A commonly used measure of beam width at focal point is defined by the following equation.[8]

$$\text{Beam width:} \quad W_b \approx f\lambda. \tag{1}$$

The measured beam width is in approximate agreement with the theoretically predicted width of 26 μm.

An excised normal rabbit eyeball was imaged with this 60 MHz self-focused needle transducer as part of an ultrasonic

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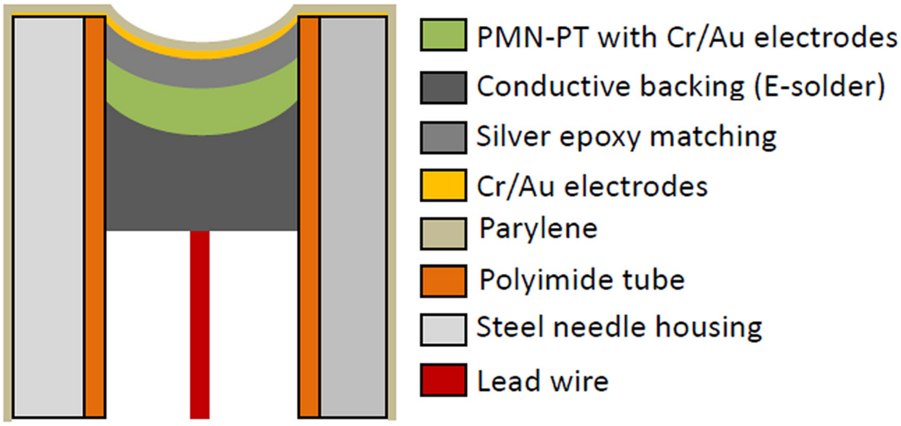

FIG. 1. Design cross section of a self-focused needle transducer.

bio-microscope (UBM) system (Fig. 3). The set-up of the UBM system was based on the design previously described by Turnbull *et al.*[9] A linear mechanical scan of the transducer was used in the system. A logarithmic compression algorithm was used to improve image grayscale visualization. One frame of image data was obtained by collecting pulse-echo lines spacing at 10 μm. The depth of image was determined by the ultrasound penetration depth, which was set to be 4 mm to image the anterior chamber of the rabbit eye. The final image was formed by converting the scan line data to 256 gray levels. The RF data used to form this image were logarithmically compressed to a 55 dB dynamic range. The image was not obtained at the focus due to limitations of the pulser of the UBM system, since the focus of the transducer (0.4 mm) actually fell in the range of the pulse (0.8 mm). Further improvement of UBM system electronics should enable us to obtain an image at the focus with better spatial resolution.

The experimental configuration of acoustic trapping is shown in Fig. 4 where the self-focused needle transducer was used in a microparticle immobilization experiment. Specifically, the microparticles were suspended in a designed chamber filled with water, sitting on top of a microstage. The chamber had an opening with an acoustically transparent mylar film at its bottom, through which particle motions were detected by an inverted microscope. The transducer was mounted on a three-axis motorized linear stage and interrogated the microparticles from above. The position of the transducer could be controlled by a computer with a customized LABVIEW code. The transducer was driven in a sinusoidal burst mode whose waveforms were generated from a function generator and then amplified by a 50 dB power amplifier to achieve desirable peak-to-peak voltage amplitude. The pulse repetition period was set to 1 ms and its duty factor was 0.03%. The trapping of the microparticles was observed through a CMOS camera attached to the microscope, and the images as well as videos captured by the CMOS camera were recorded with a computer.

To perform microparticle manipulation, polystyrene microspheres of 15 μm mean diameter were loaded into the chamber as targeted particles to be trapped. The maximum displacement of trapped particles[10] was measured and quantified to determine how far the trapping force could attract the particle from the center of the trap by this self-focused needle transducer. As shown in Fig. 5, the measurement was carried out under various excitation conditions. The frequency of the transducer was varied from 52.5 to 62.5 MHz and its peak-to-peak voltages were applied at 19, 25, and 32 Vpp. Generally, at each voltage, the displacement reached its maximum value when operating at resonance frequency of 57.5 MHz where the maximum peak pressure occurs. Meanwhile, the displacements were increased with the higher excitation voltages. The above phenomena were also observed and described in previous lipid droplets trapping experiments.[10] While we have previously demonstrated that the single beam acoustic trapping in Mie regime,[4,10] where the trapped particle size (D) is larger or close to λ, it is worth noting that the trapped particle size (15 μm) is smaller than the ultrasonic wavelength (26 μm) in this study. This phenomenon inspires us that the Rayleigh particles (D $\ll$ λ) trapping could be realized using single beam acoustic transducers, the detailed mechanism will be studied in the future work. Even with the diameter of particle, strictly speaking,

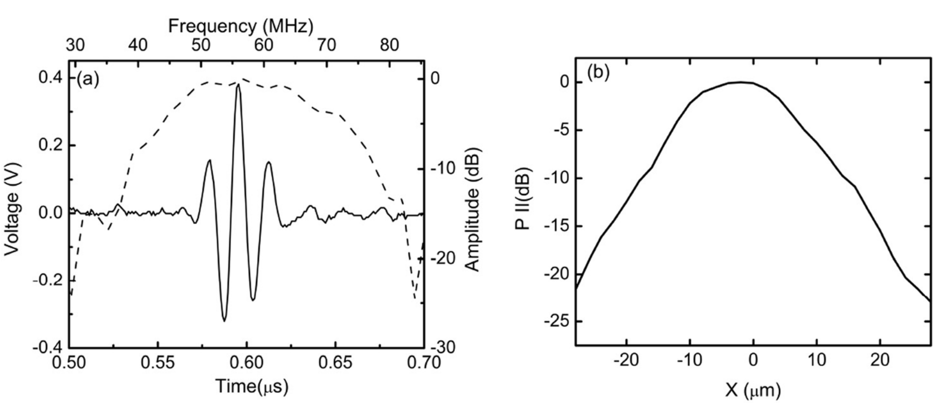

FIG. 2. (a) Measured pulse-echo waveform (solid line) and normalized frequency spectrum (dashed line). (b) Measured lateral beam profile.

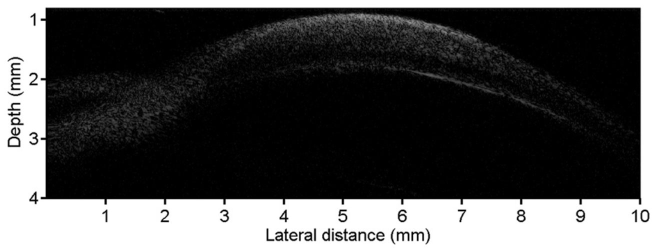

FIG. 3. *In vitro* UBM image of the anterior portion of a rabbit eye.

still being in Mie region, these experimental results indicate that single beam acoustic tweezers are capable of manipulating particles of a size not only larger but also smaller than wavelength. In case of optical tweezers, Ashkin *et al*. had reported the ability of trapping occurred over full size range from Mie to Rayleigh particles.[11] The observation of current single beam acoustic trapping experiments confirms similar capability can be achieved by acoustic tweezers as well.

In summary, using press-focusing technique, we have fabricated a miniature self-focused needle transducer (<1 mm) with a 0.4 mm focus and *f*-number equal to 1. The measured center frequency is 57.5 MHz with a bandwidth of 54% and insertion loss is 14 dB. *In vitro* UBM image of the rabbit eye has been obtained using this self-focused needle transducer. The results on polystyrene microparticle trapping by this type of transducer are exciting, since it shows that it is possible to acoustically trap particles with a size smaller than the ultrasonic wavelength. The results indicate the Rayleigh particles trapping may be feasible by single beam acoustic transducers.

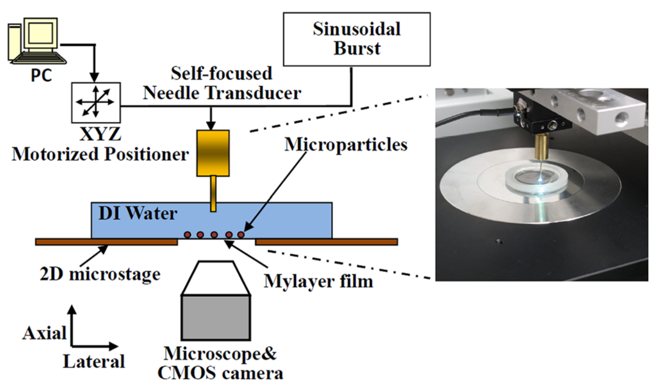

FIG. 4. Block diagram of acoustic trapping experiment using the self-focused needle transducer.

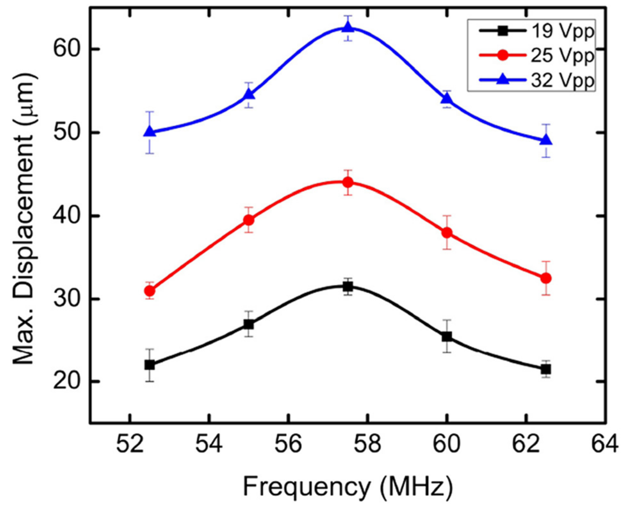

FIG. 5. Measured maximum displacements as a function of frequency with different excitation voltages.

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